Neuromimetic Event-Based Detection for Closed-Loop Tactile Feedback Control of Upper Limb Prostheses

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Abstract—Upper limb amputees lack the valuable tactile sensing that helps provide context about the surrounding environment. Here, we utilize tactile information to provide active touch feedback to a prosthetic hand. First, we developed fingertip tactile sensors for producing biomimetic spiking responses for monitoring contact, release, and slip of an object grasped by a prosthetic hand. We convert the sensor output into pulses, mimicking the rapid and slowly adapting spiking responses of receptor afferents found in the human body. Second, we designed and implemented two neuromimetic event-based algorithms, *Compliant Grasping* and *Slip Prevention*, on a prosthesis to create a local closed-loop tactile feedback control system (i.e., tactile information is sent to the prosthesis). Grasping experiments were designed to assess the benefit of this biologically inspired neuromimetic tactile feedback to a prosthesis. Results from able-bodied and amputee subjects show the average number of objects that broke or slipped during grasping decreased by over 50 percent and the average time to complete a grasping task decreased by at least 10 percent for most trials when comparing neuromimetic tactile feedback with no feedback on a prosthesis. Our neuromimetic method of closed-loop tactile sensing is a novel approach to improving the function of upper limb prostheses.

Index Terms—Neuromimetic, prosthetic, force feedback, real-time control

1 Introduction

Prosthetic hands are important tools for improving the lives of upper limb amputees; however, most of these devices lack the ability to determine and understand the sense of touch. This lack of tactile feedback can cause issues such as unstable grasping of objects as many amputees are forced to rely primarily on visual feedback to ensure their prosthetic limb is behaving appropriately. Relying primarily on visual feedback with no tactile input can be rather burdensome for an amputee when it comes to picking up, holding, or manipulating objects with their prosthesis. In healthy hands, numerous mechanoreceptors within the skin allow for our sense of touch and make up the closed-loop tactile feedback system that provides us with valuable information regarding our environment [1], [2].

Many of the prosthetic arms today are controlled using myoelectric (EMG) signals [3], [4], [5], [6]. Recent advances in EMG prosthesis control have allowed for functional improvements [7], and new EMG pattern recognition techniques have shown promise for a more natural control of a prosthesis [8], [9]. These EMG control methods are useful for creating prosthetic systems with more intuitive control, but amputees still face the problem of no tactile feedback in their control strategies. This lack of touch information can give rise to issues such as accidentally breaking or dropping an object as the prosthesis user is unable to determine the amount of grip force being used or when the object comes into contact with the prosthesis.

Knowledge gained through active touch sensing plays an important role in many manipulation tasks [10], [11], [12], [13], and research suggests that using information such as grip force and pressure can help improve the functionality and grasping control of prosthetic hands [14], [15], [16], [17], [18], [19]. Advancements in closed-loop prosthesis control include improving grasp force sensitivity by incorporating force-derivative feedback in a prosthetic hand to help regulate grasping force [17], a nonlinear force controller for estimating and reducing the force fluctuations during grasping [20], and even a hybrid force-velocity sliding mode controller for preventing excessive grasping force [21]. Recent progress has shown the benefit of providing visual force feedback to prevent slip during grasping [22] as well as using an adaptive sliding mode prosthesis control to help prevent grasped object slip and deformation [18].

Current approaches fail to take into account the biological aspects of tactile sensing, specifically the behavior of mechanoreceptors in identifying onset and offset of object contact. This type of behavior is vital for stable grasping as we rely heavily on active touch sensing to gain information

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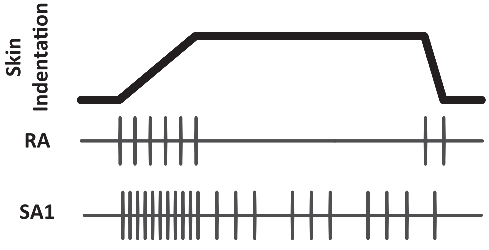

Fig. 1. Adaptation of results from [25], this schematic shows the amount of skin indentation (top) and typical RA (middle) and SA1 (bottom) responses. RA receptors respond during the transient periods of indentation to help indicate contact and release while SA1 receptors exhibit a response during sustained indentation.

of an object [23], [24]. We use the mechanoreceptors for active touch sensing as a means to better understand a task, whether that is holding an object or discriminating a fine texture [24]. In healthy skin, the transient behavior of rapidly adapting (RA) receptors is believed to send information to the peripheral nervous system regarding the onset of object contact and release while the sustained response of the slowly adapting type 1 (SA1) receptors is thought to convey information regarding the amount of static grip force (Fig. 1) [23], [25]. It has been shown that by using these event-based responses along with numerous other inputs we are able to manipulate and grab objects with high precision and reliability [1]. Drawing inspiration from biology, an event-based approach could be engineered as a neuromimetic system to enable active touch sensing for a prosthesis that relies on object contact and release events made evident through spiking behavior.

Neuromimetic systems aim to imitate some aspect of brain function using analogous neural elements, such as spiking activity [26], [27]. Neurologically inspired approaches have been employed for visual information processing [28] and object recognition [29] as well as for modeling neural circuits, eye movements, and other sensory systems [30], [31], [32], [33]. Here we model a tactile sensing system using RA 'event-based' responses to determine object contact and slip in conjunction with SA1 type information of sustained grasping force to create a neuromimetic control method for a prosthetic system. We hypothesize that this bioinspired approach will create a closed-loop tactile feedback system that can prevent object damage and slip during grasping with a prosthesis.

One study showed how different feedback modalities can influence a person's ability to detect and correct for an object slipping. The response time in healthy humans for preventing object slip by using EMG signals ranged from 1.51–1.75 s, depending on the feedback modality [34]. However, during grasping, the natural reflex pathway in healthy adults is capable of responding between 50 - 70 ms after the onset of slip occurs [35], [36]. The disconnect between muscle contractions and prosthesis movements for an upper limb amputee introduces an inherent delay when compared to the automatic skeletal muscle response triggered by efferent nerve fibers in the peripheral nervous system [2], [35], [37]. Thus, there is a need for a closed-loop tactile feedback system for prostheses with the ability to make quick, accurate adjustments in real-time during grasping, similar to our very own reflex pathway.

In this work we 1) present compliant force sensors to monitor grasping forces as active tactile sensory inputs to a

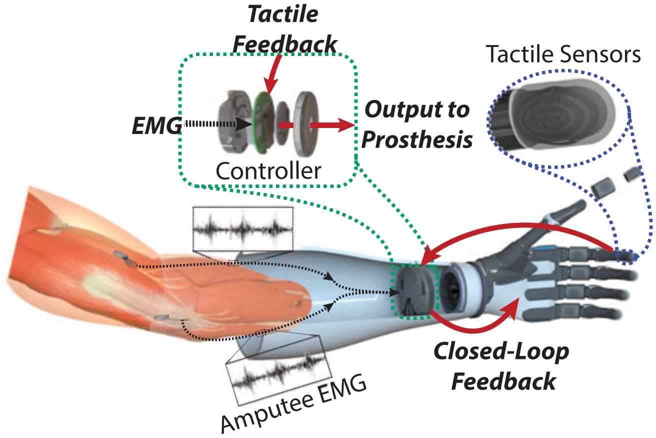

Fig. 2. System diagram showing the closed-loop nature of the tactile feedback system. The prosthesis control unit receives both amputee EMG signals and tactile information before sending out a command to the terminal device.

prosthesis control unit, and 2) implement our neuromimetic force based control algorithms, *Compliant Grasping* and *Slip Prevention*, on the prosthesis controller to create an active closed-loop tactile feedback system for improving grasping functionality of a prosthetic hand, as outlined by Fig. 2. In this work, tactile feedback is sent directly to the prosthesis controller and not the user.

2 Materials and Models

In this work we use a prosthesis control unit (Infinite Biomedical Technologies, Baltimore, USA), housed within the prosthesis socket, to interface with a bebionic3 prosthetic hand (Steeper, UK) via a standard coaxial plug. The prosthetic hand is operated using a two site EMG control strategy. The compliant fingertip sensors serve as an input into the neuromimetic algorithms, which are embedded on the control unit, to create the closed-loop tactile feedback system (Fig. 2).

2.1 Textile Force Sensor

We have designed and built a customized textile force sensitive resistor (FSR) to measure applied loads during grasping with a prosthetic hand. The sensors are based on previous designs with stretchable textiles [38] and designed specifically for the fingertips of a prosthetic hand. Sensor cuffs, as seen in Fig. 3a, are made up of stretchable conductive textile traces (LessEMF, Latham, USA) placed on a textile backing

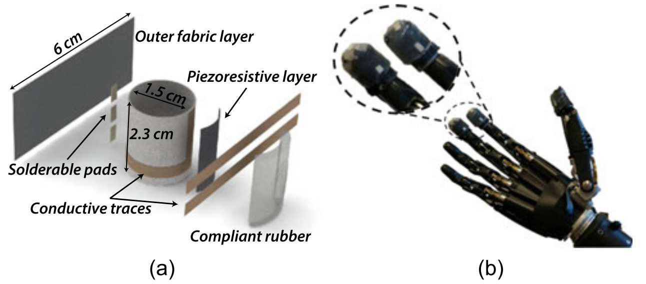

Fig. 3. (a) Textile sensor cuff design, which includes flexible and stretchable materials that allow the sensor to be placed on a prosthesis phalanx. Conductive traces act as the sensing elements and are protected by an outer fabric layer along with a rubber coating. (b) Sensor cuffs are placed on the tips of the thumb, index, and middle fingers of the prosthesis.

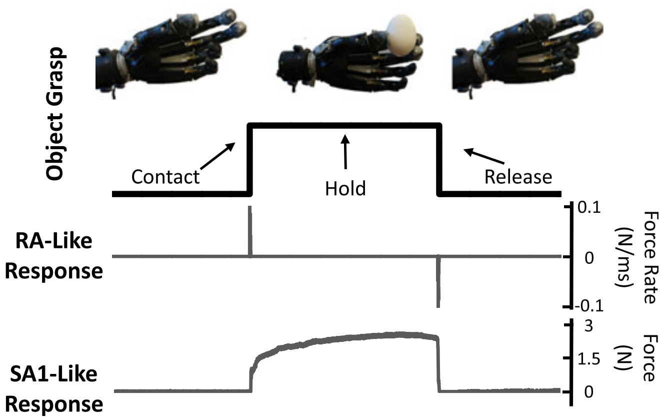

Fig. 4. A grasp-hold-release event with tactile feedback. The top plot shows the onset, hold, and release of an object grasped by a prosthetic hand. The RA-like tactile response (middle) produces a small cluster of positive spikes during the onset of object contact and negative spikes during object release. The SA1-like response (bottom) simultaneously measures sustained grip force.

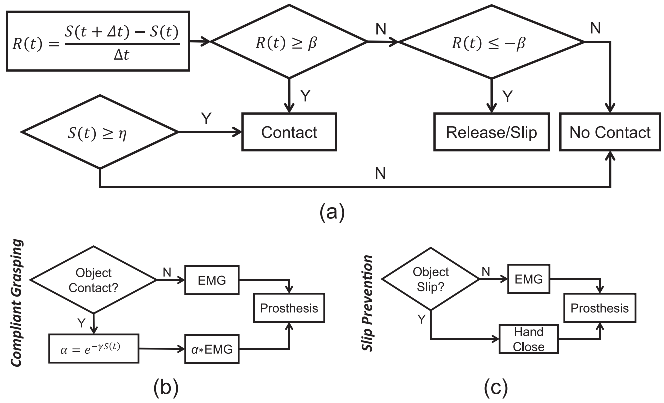

Fig. 5. (a) The neuromimetic touch feedback algorithm uses the RA-like sensor response, $R(t)$, which is found by passing the force signal ($S(t)$) through a high pass filter and comparing it to the threshold β, to determine the onset of object contact, release, and slip. (b) The *Compliant Grasping* strategy uses object contact to dynamically modulating the user's EMG gain, α, to help prevent grasping objects with excessive force, and (c) uses the same neuromimetic RA-like response to monitor and correct for object slip.

and covered by a stretchable outer layer. A 3 mm rubber layer (Dragon Skin 10, Smooth-On, USA) is used to add compliance to the grasping surface of the prosthesis. The sensors were previously characterized and verified for use in prosthetic applications, particularly during grasping tasks [39].

The textile FSRs are designed to easily fit on the phalanx of an existing prosthesis, removing any need for special disassembly or mechanical manipulation of the device. For this work, the sensor cuff (Fig. 3a) is placed on the thumb, index, and middle distal phalanges of the bebionic3 prosthetic hand, as shown in Fig. 3b. The relationship between the applied surface load and the sensor output is described in [39].

2.2 Neuromimetic Algorithms

The normal force measured by the sensors is used as an analog to skin indentations that produce RA and SA1 responses. Rapid changes in the applied force, as measured by the sensors, are translated to an RA-like spiking response to indicate object contact and release, as seen in Fig. 4. This is achieved by measuring the rate of change of the force signal and characterizing positive changes as the onset of object contact and negative changes as object release. In addition, the absolute value of a sustained applied load is simultaneously measured by the sensor to capture SA1-like information of a steady-state force or indention (Fig. 4). In our approach, these signals serve as the active tactile inputs for the neuromimetic prosthesis grasping algorithms.

2.2.1 Compliant Grasping Control

This control strategy determines when the prosthetic hand contacts an object and modulates the hand's response to the user's EMG signal during a grasping task based on the applied force from the fingertip sensors to promote a stable prosthesis grip without overexerting forces on an object. Our approach to compliant grasping is to create a device to implement a feed-forward EMG gain control model that uses an RA-like sensor response, $R(t)$, to determine object contact and the static SA1-like sensor response, $S(t)$, to determine the absolute grip force. The RA-like sensor response, $R(t)$, is modeled as a high pass filtered signal of the SA1-like sensor response, $S(t)$, and approximated using Newton's quotient

$$R(t) = \frac{S(t+\Delta t) - S(t)}{\Delta t},$$

where Δt is the time between measurements. This creates the spiking response that can be used for determining the onset of object contact and release. Object contact is defined as a threshold crossing by the RA or SA1-like sensor response

$$R(t) \geq \beta \qquad \text{or} \qquad S(t) \geq \eta,$$

with $\beta = 0.08\,\text{N/ms}$ and $\eta = 0.1\,\text{N}$, which were found experimentally to be outside of the normal force rate fluctuations of the sensor and the minimum force needed for sensor activation, respectively [39]. After object contact during a grasping task, the prosthesis control unit actively modulates the user's EMG signals by applying a gain reduction, α, which is dependent on the SA1-like sensor response, $S(t)$. This is outlined in Fig. 5 and is given by the piecewise function

$$\alpha = \begin{cases} e^{-\gamma S(t)} & S(t) < 8\text{ N} \\ \gamma & S(t) \geq 8\text{ N}, \end{cases}$$

where γ is the EMG gain threshold of 20 percent. To find γ we took the average EMG amplitude of several maximum effort contractions and found the percentage of the signal needed to maintain prosthesis control. The 8 N threshold was chosen to ensure continuity of the gain reduction function, α, as it is the intersection of the two parts of the piecewise function. Because the prosthetic hand operates using proportional control, a reduced EMG signal will result in an appropriately reduced hand reaction. The exponential decrease of the EMG gain was found heuristically to allow for finer manipulation with smaller grasping forces, which makes it ideal when handling delicate objects that are easily crushed, compared to an inversely proportional or inverse sigmoidal decaying function. Fig. 6 shows the actual EMG gain output from the prosthesis controller based on the measured force signal during active *Compliant Grasping*

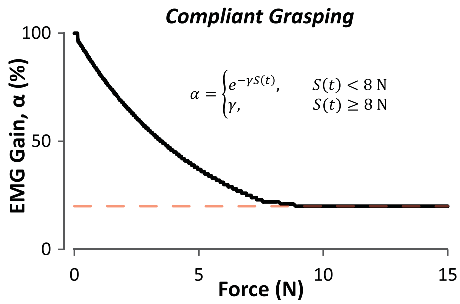

Fig. 6. The true EMG gain measured from the prosthesis controller during a prosthesis grasping task with increasing grip force and *Compliant Grasping*. To prevent the EMG signal from shrinking to zero, a lower threshold of 20 percent is placed on the gain.

feedback control. The goal of this algorithm is to allow the user to make fine force adjustments, due to the decreasing EMG gain, after contacting an object without the worry of over grasping and breaking the object.

2.2.2 Slip Prevention Control

During prosthesis grasping it is useful to have a stable grip on the target object. For this case, we introduce a neuromimetic *Slip Prevention* algorithm that uses the RA-like sensor response, $R(t)$, to determine the offset (i.e., slip) of object contact. While object contact is determined by a positive increase in $R(t)$, as described in the previous section, object release is determined by a *negative* change in $R(t)$.

$$\begin{aligned} R(t) \geq \beta \quad &\Rightarrow \quad \text{Object Contact} \\ R(t) \leq -\beta \quad &\Rightarrow \quad \text{Object Release.} \end{aligned}$$

A negative change in the grip force, less than $-\beta$, indicates movement or release between the prosthesis and grasped object interface and triggers the prosthesis to close for time τ. The value of τ is chosen as 45 ms, which is similar to actual grip force adjustment times found in humans [35]. This time was also verified experimentally as enough time for the prosthesis motors to respond to the hand close signal. The algorithm is continuously monitoring $S(t)$, so the total time of hand closure, T, is increased by τ for every instance of slip, n, and can be modeled using the update equation

$$T^i = T^{i-1} - \Delta t + \sum_{j=1}^{n} \tau_j^i,$$

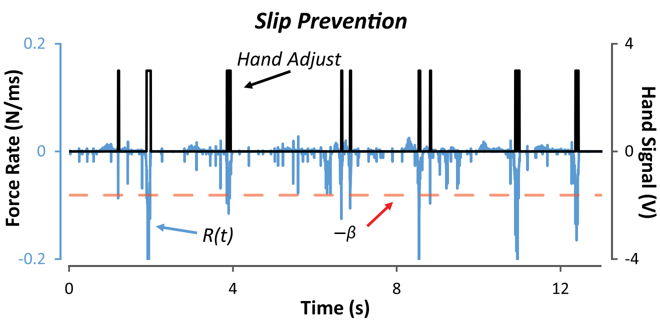

Fig. 7. The *Slip Prevention* control strategy uses the biomimetic RA-like sensor response, $R(t)$, spikes to monitor for object slip. Instances of slip are identified using this neuromimetic approach by measuring the rate of change of the grip force. An instance of slip triggers the prosthesis to close to prevent an object from slipping from its grasp.

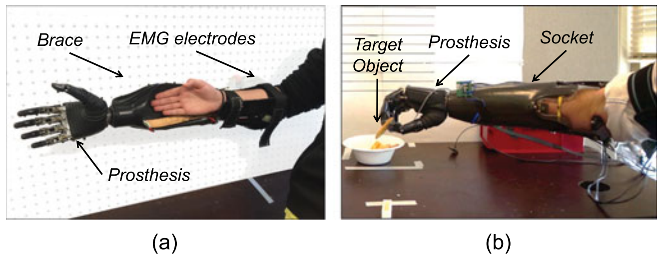

Fig. 8. (a) A custom brace is used for operation of a prosthetic hand by able-bodied subjects. A pair of Ottobock electrodes (Myobock, Ottobock, Austin, USA) are placed on the forearm of the subject to collect the EMG signals. (b) The amputee participants used their personal prosthetic socket with embedded Ottobock EMG electrodes.

where Δt is the elapsed time between iterations, i, and is dependent on the prosthesis control unit sampling rate. The prosthesis receives a close signal for time T^i, which will increase with increasing instances of slip, n. This algorithm is outlined in Fig. 5, and its output is shown in Fig. 7, which portrays the RA-like sensor response, $R(t)$ and the corresponding signal to close the prosthesis.

In the event that the user intends to release a grasped object from the prosthesis an intentional EMG "open" signal will override the automatic hand closure reflex signal from the *Slip Prevention* algorithm.

3 Experimental Methods

To evaluate the use of active tactile feedback during prosthesis operation, we developed a series of grasping experiments that require a human subject to pick up and handle objects with a bebionic3 prosthetic hand. To evaluate the algorithms with an adequate sample size, 10 able-bodied subjects participated in the experiments. To evaluate the touch feedback system with actual prosthesis users, 2 transradial amputees, one of whom is a bilateral amputee and the other a unilateral amputee, participated in experiments. All subjects consented to participate in the experiments, which were approved by the Johns Hopkins Medicine Institutional Review Board.

3.1 Hardware and Data Collection

To operate the prosthesis, able-bodied subjects wore a customized brace, Fig. 8a, while the amputee participants used their personal prosthetic sockets, Fig. 8b. A tripod grip (Fig. 9) was used by all subjects during the grasping tasks and the EMG signals used to control the prosthesis were collected using a pair of Ottobock electrodes (Myobock, Ottobock, Austin, USA) placed on the forearm of the subject. The same pair of electrodes was used for all able-bodied subjects and the amputee subjects used personal Ottobock EMG electrodes that were already embedded within their socket.

Fig. 9. A tripod grip is used by the prosthesis for all grasping tasks. For this grip, the thumb as well as the index and middle fingers are used to grasp an object.

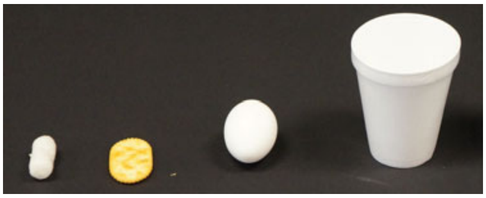

Fig. 10. The items used for the *Compliant Grasping* task. From left to right: packing foam, cracker, hollow egg, and a polystyrene cup. These common objects, most of which have been used in previous grasping studies, were chosen due to their delicate nature [14], [40], [41].

Fingertip sensors were placed on the thumb, index, and middle fingers of the prosthesis, as seen in Fig. 3b. Each sensor contains a sensing element at the distal end and tip of the finger, and they communicate directly with the prosthesis controller (Infinite Biomedical Technologies, USA) at 260 Hz. To test the individual touch feedback strategies, an external switch was placed on the prosthesis controller to change between the *Compliant Grasping* and the *Slip Prevention* algorithms. This allows each algorithm to be evaluated independently of the other. Data were sent via serial communication between the prosthesis controller and a PC and analyzed using LabVIEW (National Instruments, USA). Every experiment was recorded using a Sony Nex-5R digital camera for monitoring time and object movement during the experiments. A paired t-test with confidence interval (COI) of 95 percent is used for analyzing whether the data reject the null hypothesis when compared to each other.

3.2 Able-Bodied Experiments

Two different tasks were designed to test the functionality of the *Compliant Grasping* and *Slip Prevention* algorithms. Each able-bodied subject performed the tasks using 1) an unmodified bebionic3 prosthetic hand, 2) the prosthesis with the neuromimetic feedback for compliant grasping and slip control, and finally 3) the prosthesis with the finger sensors deactivated. There is no cosmesis, a skin-like glove, on the unmodified prosthesis. The reason for the final case with deactivated sensors is to investigate the effect of the sensors' material on the system's performance during grasping. Each subject was allowed to train with the prosthesis, both unmodified and with sensors attached, for up to a total of 15 minutes to learn basic operation and control of the device before starting the experiments.

3.2.1 Compliant Grasping

Common objects that are relatively easy to break were chosen for this task and are shown in Fig. 10. Most of these items have been used in previous prosthesis and robotic grasping tasks [14], [40], [41]. To ensure repeatability, we quantified the mass of the items as well as the amount of force required to break each item, as seen in Table 1. For this experiment, an object is considered broken when it exceeds its yield strength and undergoes plastic deformation [42]. Each trial consists of picking up and moving five items of the same type approximately 25 cm. Every able-bodied subject completed a single trial of five movements for each object type. The trials were repeated for the unmodified prosthesis, the prosthesis with the neuromimetic touch feedback algorithms, and the prosthesis with deactivated sensors. The order of trials was randomized and the number of broken objects as well as the time to complete a trial were measured.

TABLE 1
Items Used in the Grasping Tasks

Item	Mass (g)	Force to Break (N)
Foam	0.19 ± 0.01	>1
Cracker	3.1 ± 0.1	>2
Cup	3.2 ± 0.1	>2
Egg	5.7 ± 0.7	>8

3.2.2 Slip Prevention

To induce slip, weight is added to an empty polypropylene cylinder, held by the subject, in either 1 N increments (up to 5 N) or a single increment of 3.8 N. The two methods of weight addition allow for measuring the effect of small (1 N) and large (3.8 N) changes in the grasped object's weight. The weights are dropped from the top of the cylinder every time and fall approximately 12 cm to the bottom of the cylinder. The vertical distance moved by the grasped cylinder and the number of times it slipped completely from the prosthesis' grasp were measured using the high definition digital video camera at 30 fps. Each able-bodied subject performed each weight addition trial three times.

3.3 Amputee Experiments

To evaluate the neuromimetic tactile feedback system with actual prosthesis users, 2 transradial amputees participated in the experiments. Both amputee subjects used their own prosthetic system, which included the socket, electrodes, a prosthesis control unit, and a bebionic3 prosthetic hand. Both subjects regularly use their bebionic3 hand without a cosmesis during daily activities and have been using a prosthesis for 3 years or more.

The amputee subjects performed the same *Compliant Grasping* and *Slip Prevention* tasks as described in 3.2.1 and 3.2.2, respectively, and did so using 1) their unmodified bebionic3 prosthesis, 2) the prosthesis with the neuromimetic feedback for compliant grasping and slip control, and finally 3) the prosthesis with the finger sensors deactivated.

For the *Compliant Grasping* task, both amputee subjects performed four trials of every object movement with the unmodified prosthesis, the prosthesis with the neuromimetic tactile feedback, and the prosthesis with deactivated sensors. For the *Slip Prevention* task, one amputee subject performed each weight addition trial 4 times. The bilateral amputee subject did not participate in this grasping task.

4 Results

The results from the grasping tasks are separated by able-bodied and amputee subjects. The data collected from the two different grasping tasks are separated in order to evaluate the two neuromimetic algorithms independently. All error bars in the following plots represent the standard error of the mean and a paired t-test is used for analyzing the statistical significance of the able-bodied subject results. A statistical analysis was not performed for the amputee subjects' results because of the small sample size, which was also the case in [7].

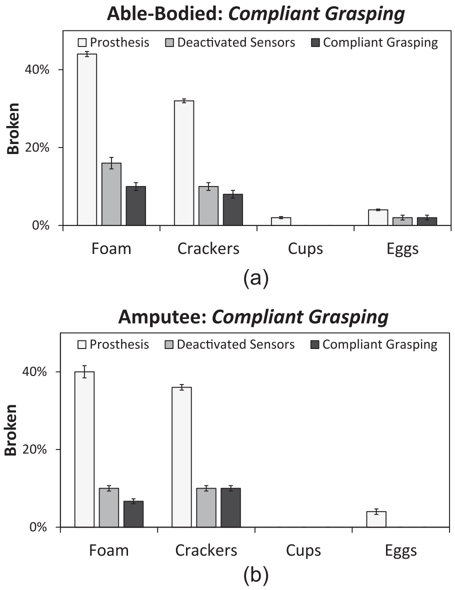

Fig. 11. The average number of broken objects during the *Compliant Grasping* tests for the (a) able-bodied and (b) amputee subjects.

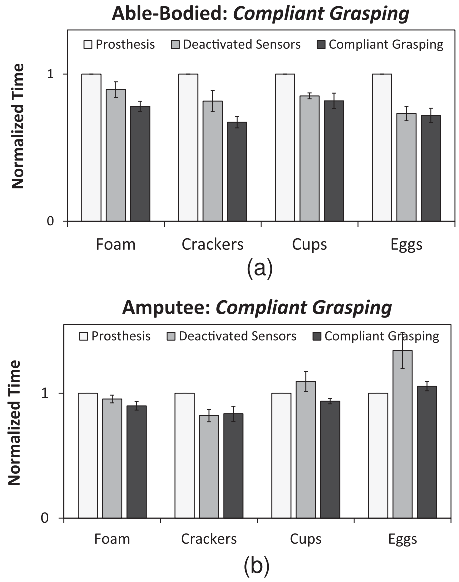

Fig. 12. The normalized time to complete a *Compliant Grasping* tests for the (a) able-bodied and (b) amputee subjects. Trial completion times are normalized using the average time to complete a task for a particular item using the unmodified prosthesis. Both plots show a decrease in the time required to complete item movements while using tactile feedback as an input for the control algorithm, with the exception of the eggs for the amputee subjects.

4.1 Compliant Grasping

For both subject types, the average number of broken items, as a percentage of the total number of items moved, is shown in Fig. 11. An object is considered broken if its elastic limit is exceeded and it undergoes plastic deformation during the grasp [42]. In general, the unmodified prosthesis broke the most number of items, and the number of broken objects decreases significantly with the use of compliant fingertip sensors and feedback. Fig. 12 shows the normalized time for completing a trial based on the target object. To allow comparison across the subjects of a group, the time to complete a trial for an object was normalized against the completion time of using the unmodified prosthesis for that same object.

4.1.1 Able-Bodied Subjects

The number of broken objects (Fig. 11a) dropped from 44, 32, 2, and 4 percent while using the unmodified prosthesis to 16, 10, 0, and 2 percent while using the prosthesis with deactivated sensors for the foam pieces, crackers, cups, and eggs, respectively, for able-bodied subjects. The failure rate for the foam and crackers decreased further to 10 and 8 percent while there was no change for the cups and eggs with the *Compliant Grasping* algorithm. There is a statistical significance ($p < 0.05$) between the results from the unmodified prosthesis and those from the neuromimetic closed-loop tactile feedback. There is a statistical significance observed between results from the deactivated sensors and those with the tactile feedback for the foam ($p = 0.01$) and crackers ($p = 0.04$) but not for the other two items.

Using the unmodified prosthesis resulted in the longest trial completion times for the able-bodied subjects. The normalized completion time changed from 0.89, 0.82, 0.85, and 0.73 with the deactivated sensors to 0.78, 0.67, 0.82, and 0.72 while using the neuromimetic touch feedback for the foam, crackers, cups, and eggs, respectively (Fig. 11a). There is a statistically significant difference ($p < 0.05$) between results from using the prosthesis and the *Compliant Grasping* algorithm. This is also true for the results from the deactivated sensors and the tactile feedback for the foam and crackers but not for the cups ($p = 0.19$) or the eggs ($p = 0.72$).

4.1.2 Amputee Subjects

The number of broken objects decreased from 40, 36, and 4 percent while using a prosthesis to 10, 10, and 0 percent while using a prosthesis with deactivated sensors to grab the foam, crackers, and eggs, respectively, for the amputee subjects (Fig. 11b). Utilizing the tactile feedback with the *Compliant Grasping* algorithm further decreased the broken foam to 7 percent while the number of broken crackers and eggs stayed the same. No cups were broken by the amputees on any trial.

The normalized completion times while using the prosthesis with the deactivated sensors are 0.95, 0.82, 1.09, and 1.34 for the foam, crackers, cups, and eggs, respectively. These times are reduced to 0.90, 0.84, 0.94, and 1.06 while using active touch feedback control, as seen in Fig. 12b .

4.2 Slip Prevention

The average distance slipped by the cylinder during the small (1 N) or large (3.8 N) weight addition for the *Slip Prevention* grasping task was measured and is shown in Fig. 13. All instances of slip were less than 1 s in duration. Fig. 14 shows the failed trials, which are defined as the cylinder slipping entirely from the grasp of the prosthesis during weight addition.

4.2.1 Able-Bodied Subjects

The average distance slipped while using a prosthesis is 8.3 mm and 25.5 mm for the small (1 N) and large (3.8 N) weight additions, respectively. These values are reduced to 1.2 mm and 6.1 mm while using deactivated sensors on the prosthesis and further reduced to 0.8 mm and 3.8 mm while

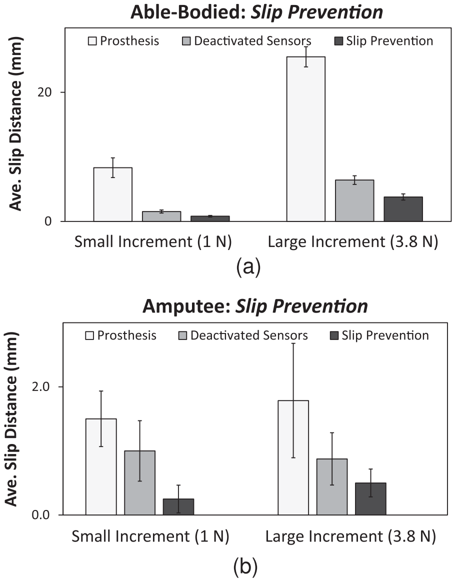

Fig. 13. The average distance the grasped cylinder slipped during the *Slip Prevention* tests for the (a) able-bodied and (b) amputee subjects.

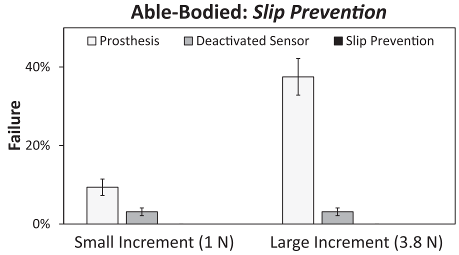

Fig. 14. The number of times, as a percentage of the total number of trials, the grasped cylinder fell from the prosthesis during the *Slip Prevention* tests for the able-bodied subjects. There were no failed trials during experiments with the amputee subject.

using the *Slip Prevention* algorithm (Fig. 13a). 9 percent of the trials resulted in complete slip (i.e., failure) while using the prosthesis during small weight increments and 38 percent for the large weight increment. Both of these failure rates reduced to 3 percent with the presence of the deactivated sensors and were reduced even further to 0 percent with the neuromimetic algorithm to prevent slip (Fig. 14). There is a statistically significant result ($p < 0.05$) between all trials with the unmodified prosthesis and those utilizing the *Slip Prevention* algorithm; however, this is not the case when comparing the results from the prosthesis with deactivated sensors and *Slip Prevention*. The resulting p values for this comparison (deactivated sensors versus *Slip Prevention*) from the slip distance data for the small and large weight increments are 0.24 and 0.19, respectively.

4.2.2 Amputee Subject

While using the unmodified prosthesis, small and large weight additions resulted in an average slip distance of 1.5 mm and 1.8 mm, respectively, for the amputee subject. The presence of the deactivated sensors reduced the slip distance to 1.0 mm and 0.9 mm for the small and large weight additions, respectively, while using the neuromimetic *Slip Prevention* resulted in distances of 0.3 mm and 0.5 mm for the same weight increments (Fig. 13a). There were no instances of failed trials during this task.

5 Discussion

This system is the first to incorporate active neuromimetic touch feedback algorithms on a prosthetic hand. Using event-based spiking activity from force sensors on the fingertips of the prosthesis, results from the *Compliant Grasping* and *Slip Prevention* algorithms suggest the benefit of using such an event-based approach for closed-loop control of a prosthetic hand.

5.1 Compliant Grasping

In general, the *Compliant Grasping* control strategy appears to benefit the user by reducing the number of broken objects during grasping. From Fig. 11, it is clear that the lack of tactile feedback makes it difficult to grab delicate objects without breaking them, which has been seen in other studies as well [13], [17], [22], while the presence of the neuromimetic tactile feedback system reduces the likelihood of objects breaking. Interestingly, the deactivated sensors also reduced the number of broken items compared to the unmodified prosthesis, but to a lesser degree. This is likely caused by the compliant nature of the sensor surface, thus helping distribute any grasping loads over a larger surface area and in turn reducing the pressure applied to an object during grasping.

5.1.1 Able-Bodied Subjects

The reduction of broken objects when using the deactivated fingertip sensors (Fig. 11a) highlights the effect of the compliant sensor surfaces on the grasping performance of the prosthesis; however, the presence of the active neuromimetic tactile feedback showed even better performance, specifically for the more delicate objects. Similarly with the normalized completion time, the *Compliant Grasping* algorithm has the best performance (Fig. 12a). This enhanced performance can be attributed to the RA-like event-based responses for determining the onset of object contact. The knowledge of initial contact allows the tactile feedback system to effectively reduce the prosthesis grasping force through EMG modulation, which is a function of the SA1-like sensor response as described in 2.2.1.

The likelihood of breaking delicate objects during prosthesis grasping was reduced by using a force derivative feedback in [17] as well as visual force feedback in [22]; however, this neuromimetic tactile feedback approach results in higher success rates for grasping delicate objects.

5.1.2 Amputee Subjects

Similarly, there is a reduction in failure rates when handling the objects with the deactivated sensors on the prosthesis and a further reduction with the neuromimetic algorithm (Fig. 11b). It is interesting to note that the cup was never broken by an amputee subject nor were there any failures with the closed-loop tactile feedback for the eggs. One possible reason for this is that the amputee subjects are much more experienced using their devices and are able to control it with more precision than a naive user, thus they are more capable of handling objects that require slightly higher force to break ($F > 8$ N); however, the presence of the neuromimetic tactile feedback was still beneficial to improve grasping of objects, specifically those that are very delicate (i.e., $F \leq 2$ N to break).

There are slight decreases in completion time for all manipulation tasks, except for with the egg, while using the *Compliant Grasping* feedback strategy (Fig. 12b). One amputee subject claimed that he was so comfortable operating his prosthesis in an unmodified state, without any cosmesis or covering, on a daily basis that adding the fingertip sensors caused observable changes in how he used the device, particularly in how quickly he would pick up objects. As a result, the active neuromimetic tactile feedback still provides the added benefit of reducing the likelihood of breaking objects but may not have a profound effect on the time to complete a grasp as many experienced prosthesis users are already proficient regarding the time required to grab and move an object. While this may be the case in general, we still observe reduced completion times for the three most delicate objects with the presence of the active tactile feedback.

5.2 Slip Prevention

During the *Slip Prevention* task, the spiking neuromimetic feedback allowed the system to noticeably reduce the amount of slip during weight addition to the grasped object (Fig. 13). Although the sensors do not isolate the changes in tangential loading during slip, a negative spike in $R(t)$ indicates an instance of object slip and leads to a hand reaction to stop the object from falling (Fig. 7), not unlike the healthy reflex pathway. This *Slip Prevention* method relies solely on the RA-like sensor signal to monitor and correct for rapid changes in the grip force caused by a slipping object. This is similar to actual human behavior where adjustments in grip forced are caused by changes in the vertical load of the grasped object [35]. Interestingly, it has been shown that the grip force adjustment is proportional to the magnitude of the vertical load perturbation but is not related to the preexisting grip force [35]. By drawing parallels with biology, our neuromimetic *Slip Prevention* tactile feedback algorithm has demonstrated ability to reduce object slip during grasping with a prosthesis.

Previous studies have shown the ability to prevent object slip [34], [40]. A series of experiments that tested ability of a user to produce EMG signals to stop a simulated slip showed mean user response times greater than 1.5 s with a success rate not exceeding 30 percent [34]. Although the experimental conditions differ, the neuromimetic feedback system showed greater ability in reducing slip failure rates suggesting the importance of prompt reaction times. One prosthesis experiment of 20 slip trials resulted in only 1 failure [40]. It should be noted that these experiments were performed without human subjects and with a limited range of detectable forces. Although the *Slip Prevention* algorithm didn't result in any slip failures, one possible area of investigation would be the performance of the system over a much larger range of slip conditions, such as slip speed.

5.2.1 Able-Bodied Subjects

There is a significant reduction in the distance slipped by the grasped object while using a prosthesis with the deactivated sensors; however, there is a further reduction with the *Slip Prevention* feedback control (Fig. 13a). The compliant nature of the sensors appears to have the a major impact on preventing slip. This is likely due to the increased surface area and friction introduced by the compliant material of the sensor itself; however, the closed-loop *Slip Prevention* algorithm is still beneficial, although not significantly so ($p > 0.05$), in that it is able to further reduce the amount of object slip by monitoring the $R(t)$ sensor signal for instances of grip force perturbations. More importantly, the active touch feedback successfully prevented any instant of major or complete object slip (Fig. 14).

5.2.2 Amputee Subject

There is an obvious decrease in the slip distance when the deactivated sensors or on the prosthetic hand, again likely due to their larger coefficient of friction than the plastic prosthesis phalanges, and a further decrease when using the *Slip Prevention* feedback (Fig. 13b). Because of the relatively small sample size, any statistical backing of these results is unclear; however, these results follow a similar trend to those seen with the able-bodied subjects. The overall small magnitude of the slipped distance as well as the lack of any slip failures can be attributed to the amputee's natural desire to "over grasp" the cylinder. When asked about his grasping tendencies, the unilateral amputee indicated that it is common for him to naturally use a larger grip force than necessary, especially with a sturdy object such as the cylinder used in this experiment, to overcompensate for the lack of feedback that is used to prevent object slip. This being the case, small amounts of slip were still observed and were reduced when using the event-based neuromimetic algorithm for tactile feedback. This suggests that despite an amputee's best efforts to properly grasp an object there are still instances of accidental object slip, which could be mitigated with the addition of a biologically inspired neuromimetic *Slip Prevention* system.

5.3 Active Touch Sensing

The tactile feedback sent to the prosthesis controller directly influences the behavior of the limb in order to better complete the current task. Whether handling delicate objects (*Compliant Grasping*) or trying to keep grasped objects steady (*Slip Prevention*), the sensory information from the fingertips plays a key role in the decisions made by the controller. The primary goal being that the control of the prosthesis is updated to whatever manner best suits the current task. This aspect of the system is analogous to the natural behavior in healthy grasping in which rapid and reliable cues are used to control our behavior during such a task [24].

This neuromimetic tactile feedback system attempts to use active touch sensing to further improve how a prosthetic limb operates by drawing parallels to healthy biological systems, specifically our ability to reliably and comfortably manipulate and grab objects.

5.4 General Considerations

Careful consideration must be made before making any comparisons between able-bodied and amputee subjects as the two groups are inherently different, but there are a few interesting aspects to note. One is the reduction in time to complete the *Compliant Grasping* task. Able-bodied subjects show larger improvements with the addition of the automatic event-based tactile feedback algorithm; however this can likely be attributed to the fact that they are naive prosthesis users and have a larger room for improvement compared to experienced prosthesis users. Experienced prosthesis users are more likely to be efficient in terms of their ability to use the unmodified prosthesis, as discussed in 5.1.2. Amputee subjects are typically more comfortable with operating a prosthesis and so any changes to the device they are already comfortable using could result in reduced performance as the modifications are unfamiliar. Despite this possibility, the active touch feedback was still able to improve prosthesis grasping.

Another aspect between the two groups is the large difference in the object slip distance. The distance slipped during able-bodied trials tends to be an order of magnitude higher than for the amputee subject. This is likely due to the higher grasping force of the amputee, as mentioned in 5.2.2. This could also be attributed to the able-bodied subjects being naive users who are unfamiliar with efficient prosthesis grasping techniques. In fact, data from the SA1-like sensor response, $S(t)$, shows that the grasping force for these trials was indeed higher for the amputee subject than it was for the able-bodied subjects. In addition, this could potentially be attributed to the different torques produced at the distal end of the prosthesis with the addition of weight. Because the brace worn by the able-bodied subjects extends further than the user's arm, a torque is produced by the terminal device, the prosthesis, and any added weight, which creates an upward force on the arm of the subject. This could potentially cause the user to quickly stabilize his or her arm with an opposing force, which would effectively move the prosthesis upward and could allow for additional slip of the grasped object.

The physical presence of the fingertip sensors improves the prosthesis grasping functionality. The compliant nature of the sensors' surface provides increased surface area during grasping while also increasing the friction between the prosthesis and the target object. A similar effect was found in [14] where a majority of prosthesis grasping improvements were found to be linked with the compliant nature of the sensors used in the tactile feedback control. This fact is not surprising as it has been shown that the compliant nature and mechanical deformation of human finger pads work in tandem with the mechanoreceptors in the human skin and are used to enhance human grasping [43], [44], [45].

Studies have shown the benefit of vibrotactile feedback to a user for preventing object slip [13], [34]. However, there is a delay with this type of feedback before a user's reaction. Given the short time scale of this particular application ($< 1\ s$), it is necessary for direct, closed-loop feedback to the prosthesis controller in order for reaction quick enough to prevent object slip or damage. There is a possible benefit of combining our neuromimetic feedback to the prosthesis controller with feedback to the user; however, it will likely require that the feedback to the prosthesis controller trigger the primary response due to the time delay of feedback to the user before a reaction.

Grasping and the sense of touch is an extremely complicated biological system that is only a portion of the even more complicated neuromuscular system. Our active neuromimetic tactile feedback system by no means attempts to model all the neurological aspects of tactile feedback during grasping, instead our method focuses on using two key elements to convey grasping information – RA and SA1 mechanoreceptor responses. Using this as a model, we can extract meaningful information regarding the onset of object contact and release to create an event-based detection system for improving prosthesis grasping.

5.4.1 Subjective Evaluation

Both amputee subjects were interviewed after the experiments to provide feedback on the proposed method. One subject noted that there was no significant perceived difference in their ability to pick up or move objects between the various experiments. This subject did indicate that the physical presence of the sensors felt awkward in the sense that it changed the thickness of the fingertips and was different than what this subject is used to. This subject indicated that a feedback system like this could be useful if it was seamlessly integrated with the prosthetic system without affecting the user's normal operation of the device. The biggest drawback for this subject was the added thickness of the fingertips due to the presence of the sensors. The subject did agree though that the presence of the compliant tactile sensors offered a benefit for reducing broken objects. Likewise, the other subject describe a sensation of being able to "feel" the presence of the compliant sensors and their ability to reduce the number of broken objects during grasping. This subject indicated that although no feedback was given to the user, the compliant nature of the sensors and the feedback to the prosthesis appeared to make it easier while grabbing delicate objects.

6 Conclusion

Our novel approach uses RA and SA1-like sensor responses to create a neuromimetic event-based tactile feedback system, which is shown to offer improvements in grasping over a traditional open loop prosthesis system. The primary goal of this investigation is to provide tactile feedback to a prosthetic hand by drawing inspiration from nature. In doing so, we have successfully shown the added benefit of implementing neuromimetic tactile feedback algorithms, *Compliant Grasping* and *Slip Prevention*, for not only enhancing ability of prosthesis users to pick up and manipulate delicate objects but to also reduce accidental slip in objects that are being perturbed by changes in weight. This neuromimetic approach offers a new insight into the improvements that can be made towards prosthesis functionality by using natural human neurological function as a platform.

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References

[1] R. S. Johansson, "Sensory input and control of grip," in *Proc. Novartis Found. Symp.*, 1998, vol. 218, pp. 45–63.

[2] S. S. Hsiao and M. Gomez-Ramirez, "Neural mechanisms of tactile perception," in *Comprehensive Handbook of Psychology: Volume 3: Behavioral Neuroscience*, M. Gallagher and R. Nelson, Eds., 2nd ed. New York, NY, USA: John Wiley and Sons Inc, 2012.

[3] P. Parker and R. Scott, "Myoelectric control of prostheses," *Crit. Rev. Biomed. Eng.*, vol. 13, no. 4, pp. 283–310, 1986.

[4] P. Parker, K. Englehart, and B. Hudgins, "Myoelectric signal processing for control of powered limb prostheses," *J. Electromyogr. Kinesiol.*, vol. 16, no. 6, pp. 541–548, Dec. 2006.

[5] K. Englehart and B. Hudgins, "A robust, real-time control scheme for multifunction myoelectric control," *IEEE Trans. Bio Med. Eng.*, vol. 50, no. 7, pp. 848–854, Jul. 2003.

[6] A. Fougner, O. Stavdahl, P. J. Kyberd, Y. G. Losier, and P. A. Parker, "Control of upper limb prostheses: Terminology and proportional myoelectric control-a review," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 20, no. 5, pp. 663–677, Sep. 2012.

[7] S. M. Wurth and L. J. Hargrove, "A real-time comparison between direct control, sequential pattern recognition control and simultaneous pattern recognition control using a fitts' law style assessment procedure," *J. Neuroeng. Rehabil.*, vol. 11, no. 91 pp. 1–13, 2014.

[8] M. A. Powell, R. R. Kaliki, and N. V. Thakor, "User training for pattern recognition-based myoelectric prostheses: Improving phantom limb movement consistency and distinguishability," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 22, no. 3, pp. 522–532, May 2014.

[9] L. H. Smith, T. A. Kuiken, and L. J. Hargrove, "Real-time simultaneous and proportional myoelectric control using intramuscular emg," *J. Neural. Eng.*, vol. 11, no. 6, pp. 1–13, Dec. 2014.

[10] T. L. Gibo, A. J. Bastian, and A. M. Okamura, "Grip force control during virtual object interaction: Effect of force feedback, accuracy demands, and training," *IEEE Trans. Haptics*, vol. 7, no. 1, pp. 37–47, Mar. 2014.

[11] C. King, M. O. Culjat, M. L. Franco, C. E. Lewis, E. P. Dutson, W. S. Grundfest, and J. W. Bisley, "Tactile feedback induces reduced grasping force in robot-assisted surgery," *IEEE Trans. Haptics*, vol. 2, no. 2, pp. 103–110, Apr.–Jun. 2009.

[12] D. Prattichizzo, C. Pacchierotti, and G. Rosati, "Cutaneous force feedback as a sensory subtraction technique in haptics," *IEEE Trans. Haptics*, vol. 5, no. 4, pp. 289–300, Jan. 2012.

[13] J. Walker, A. Blank, P. Shewokis, and M. O'Malley, "Tactile feedback of object slip facilitates virtual object manipulation," *IEEE Trans. Haptics*, vol. 8, no. 4, pp. 454–466, Apr. 2015.

[14] B. Matulevich, G. E. Loeb, and J. A. Fishel, "Utility of contact detection reflexes in prosthetic hand control," in *Proc. Int. Conf. Intel. Robo. Syst.*, 2013, pp. 4741–4746.

[15] T. D'Alessio and R. Steindler, "Slip sensors for the control of the grasp in functional neuromuscular stimulation," *Med. Eng. Phys.*, vol. 17, no. 6, pp. 466–470, Sep. 1995.

[16] P. J. Kyberd and P. H. Chappell, "Object-slip detection during manipulation using a derived force vector," *Mechatronics*, vol. 2, no. 1, pp. 1–13, 2 1992.

[17] E. D. Engeberg and S. Meek, "Improved grasp force sensitivity for prosthetic hands through force-derivative feedback," *IEEE Trans. Bio. Med. Eng.*, vol. 55, no. 2, pp. 817–821, Feb. 2008.

[18] E. D. Engeberg and S. G. Meek, "Adaptive sliding mode control for prosthetic hands to simultaneously prevent slip and minimize deformation of grasped objects," *IEEE/ASME Trans. Mechatronics*, vol. 18, no. 1, pp. 376–385, Feb. 2013.

[19] A. Ajoudani, S. B. Godfrey, M. Bianchi, M. G. Catalano, G. Grioli, N. Tsagarakis, and A. Bicchi, "Exploring teleimpedance and tactile feedback for intuitive control of the pisa/iit softhand," *IEEE Trans. Haptics*, vol. 7, no. 2, pp. 203–215, Apr.–Jun. 2014.

[20] C. F. Pasluosta, H. Tims, and A. W. L. Chiu, "Slippage sensory feedback and nonlinear force control system for a low-cost prosthetic hand," *Am. J. Biomed. Sci.*, vol. 1, no. 4, pp. 295–302, 2009.

[21] E. D. Engeberg, S. G. Meek, and M. A. Minor, "Hybrid force-velocity sliding mode control of a prosthetic hand," *IEEE Trans. Bio. Med. Eng*, vol. 55, no. 5, pp. 1572–1581, May 2008.

[22] E. D. Engeberg and S. Meek, "Enhanced visual feedback for slip prevention with a prosthetic hand," *Prosthet. Orthot. Int*, vol. 36, no. 4, pp. 423–429, Dec. 2012.

[23] G. Westling and R. S. Johansson, "Responses in glabrous skin mechanoreceptors during precision grip in humans," *Exp. Brain Res.*, vol. 66, no. 1, pp. 128–140, 1987.

[24] T. J. Prescott, M. E. Diamond, and A. M. Wing, "Active touch sensing," *Phil. Trans. R. Soc. B.*, vol. 366, no. 1581, pp. 2989–2995, Nov. 2011.

[25] R. S. Johansson and Å. Vallbo, "Tactile sensory coding in the glabrous skin of the human hand," *Trends Neurosci.*, vol. 6, pp. 27–32, 1983.

[26] F. Grassia, L. Buhry, T. Lévi, J. Tomas, A. Destexhe, and S. Saïghi, "Tunable neuromimetic integrated system for emulating cortical neuron models," *Front. Neurosci.*, vol. 5, pp. 1–12, 2011.

[27] C. Mead, *Analog VLSI and Neural Systems*. Boston, MA, USA: Addison-Wesley Longman Publishing Co., Inc, 1989.

[28] R. J. Vogelstein, U. Mallik, E. Culurciello, G. Cauwenberghs, and R. Etienne-Cummings, "A multichip neuromorphic system for spike-based visual information processing," *Neural Comput.*, vol. 19, no. 9, pp. 2281–2300, 2007.

[29] G. Orchard, J. G. Martin, R. J. Vogelstein, and R. Etienne-Cummings, "Fast neuromimetic object recognition using fpga outperforms gpu implementations," *IEEE Trans. Neural Netw. Learn. Syst.*, vol. 24, no. 8, pp. 1239–1252, 2013.

[30] S. D. Ha, J. Shi, Y. Meroz, L. Mahadevan, and S. Ramanathan, "Neuromimetic circuits with synaptic devices based on strongly correlated electron systems," *Phys. Rev. Appl.*, vol. 2, no. 6, pp. 1–11, Dec. 2014.

[31] P. M. Daye, L. M. Optican, E. Roze, B. Gaymard, and P. Pouget, "Neuromimetic model of saccades for localizing deficits in an atypical eye-movement pathology," *J. Trans. Med.*, vol. 11, pp. 1–14, 2013.

[32] L. L. Bologna, J. Pinoteau, J.-B. Passot, J. A. Garrido, J. Vogel, E. R. Vidal, and A. Arleo, "A closed-loop neurobotic system for fine touch sensing," *J. Neural Eng.*, vol. 10, no. 4, pp. 1–17, 2013.

[33] F. Corradi, D. Zambrano, M. Raglianti, G. Passetti, C. Laschi, and G. Indiveri, "Towards a neuromorphic vestibular system," *IEEE Trans. Biomed. Circuits Syst.*, vol. 8, no. 5, pp. 669–680, Oct. 2014.

[34] D. D. Damian, A. H. Arita, H. Martinez, and R. Pfeifer, "Slip speed feedback for grip force control," *IEEE Trans. Bio. Med. Eng.*, vol. 59, no. 8, pp. 2200–2210, Aug. 2012.

[35] K. J. Cole and J. H. Abbs, "Grip force adjustments evoked by load force perturbations of a grasped object," *J. Neurophys.*, vol. 60, no. 4, pp. 1513–1522, 1988.

[36] M. De Gregorio and V. Santos, "Human grip responses to perturbations of objects during precision grip," in *The Human Hand as an Inspiration for Robot Hand Development*, vol. 95, R. Balasubramanian and V. J. Santos, Eds. Berlin, Germany: Springer International Publishing Switzerland, 2014, pp. 159–188.

[37] R. Johansson and G. Westling, "Roles of glabrous skin receptors and sensorimotor memory in automatic control of precision grip when lifting rougher or more slippery objects," *Exp. Brain Res.*, vol. 56, no. 3, pp. 550–564, 1984.

[38] W. W. Lee, J. J. Cabibihan, and N. V. Thakor, "Biomimetic strategies for tactile sensing," in *Proc. Int. Conf. Sens.*, 2013, pp. 1–4.

[39] L. Osborn, W. W. Lee, R. Kaliki, and N. V. Thakor, "Tactile feedback in upper limb prosthetic devices using flexible textile force sensors," in *Proc. Int. Conf. Bio. Med. Robo. Biomech.*, 2014, pp. 114–119.

[40] N. Wettels, A. R. Parnandi, J.-H. Moon, G. E. Loeb, and G. S. Sukhatme, "Grip control using biomimetic tactile sensing systems," *IEEE/ASME Trans. Mechatronics*, vol. 14, no. 6, pp. 718–723, Dec. 2009.

[41] J. M. Romano, K. Hsiao, G. Niemeyer, S. Chitta, and K. J. Kuchenbecker, "Human-inspired robotic grasp control with tactile sensing," *IEEE Trans. Robot.*, vol. 27, no. 6, pp. 1067–1079, Dec. 2011.

[42] L. N. Kachanov, *Plastic Deformation, Principles and Theories*, H. H. Hausner, Ed. Brooklyn: Mapleton House, 1948.

[43] K. O. Johnson and J. Phillips, "Tactile spatial resolution. III. A continuum mechanics model of skin predicting mechanoreceptor responses to bars, edges, and gratings." *J. Neurophys.*, vol. 46, no. 6, pp. 1204–1225, 1981.

[44] Q. Wang and V. Hayward, "In vivo biomechanics of the fingerpad skin under local tangential traction," *J. Biomech.*, vol. 40, no. 4, pp. 851–860, 2007.

[45] B. Delhaye, P. Lefvre, and J.-L. Thonnard, "Dynamics of fingertip contact during the onset of tangential slip," *J. R. Soc. Interface*, vol. 11, no. 100, pp. 1–11, 2014.

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